

Idiopathic scoliosis: relations between the Cobb angle and the dynamical strategies when sitting on a seesaw

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Abstract The aim of this study was to determine the influence of the severity of the spinal curve on the postural regulation when self-imposed disturbances occur in a seated position in anteroposterior (AP) and mediolateral (ML) orientations. Twelve female adolescents with a right thoracic scoliosis (Cobb = 30.4° ± 9.7) were included in this study. The ground reaction forces (GRF) were studied while the subjects were maintaining their sitting on a seesaw (ML or AP destabilisation). Five conditions were tested: eyes open; with additional loads placed onto the subject's right or left shoulder; or onto the subject's right or left pelvis. We tested the correlation between the Cobb angle and the postural parameters (index of performance and GRF variability) for each condition. When the destabilisation was AP, the Cobb angle was significantly correlated with GRF variability and anterior and concavity index of performance. Two conditions showed higher correlations: stabilisation with the concavity pelvis load (GRF variability) and the open eyes (index of performance). In contrast, whatever the condition tested was, no link was found when the destabilisations were applied in ML direction. The destabilisation in a seated position highlights the influence of the curve severity on the postural organisation. In seated position, the postural control strategies specific to the scoliotic patients were always correlated by severity of curve, especially

when the destabilisation is applied in AP directions. This study showed that the unstable seating position can be considered as a pertinent paradigm to help finding a postural clinical index for adolescent idiopathic scoliosis.

Keywords Idiopathic scoliosis · Cobb angle · Postural control · Sitting balance

Introduction

2–4% of teenagers aged between 10 and 16 suffer from adolescent idiopathic scoliosis (AIS) [13]. AIS is a progressive growth disease that affects spinal anatomy, mobility, and left-right symmetry [3, 21]. The AIS imply an asymmetrical trunk with asymmetrical posture which modify posturodynamic organisation [32]. In the processes underlying the control of scoliotic patients' postural balance, specific sensory factors may lead to the development of adaptive strategies [30]. The specific postural strategies occurring in AIS have been previously studied during upright stance [8], stepping initiation [4], walking [10] and side-stepping initiation [4]. Scoliotic patients show an increase of the amplitude of the postural oscillations [27] associated with instability [8, 25] and an increased asymmetry of the dynamics of the lower limbs [5, 18, 29]. The variability of the parameters analysed was greater for scoliotic patients than for control subjects [4], especially in the mediolateral (ML) and anteroposterior (AP) directions [16, 30]. The dynamical strategies observed have been found to result in slower movements during normal walking, walking on a beam, and lateral stepping [4, 22].

In seated position, some authors have reported that spinal deformity is associated with an increase of postural trunk stabilisation compared to healthy subjects [2, 28]. In

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contrast, Smith et al. [32] reported that the spinal curvature imply an increase of the variability and asymmetry forces. More precisely, in a sitting position, the AIS subjects showed that the lateral displacement [28], spinal muscle activity [17] and ischial thrust [32] are significantly larger on the convex side of the deformation than on the concave side. Nevertheless, in dynamical condition (seated balance and initiation step), the concavity side is systematically more disturbed than the convexity side [5, 6]. During seating self-stabilisation on a seesaw, the addition of mass onto the concavity pelvis side increased the patient imbalance [6]. Chockalingam [10] has described a special process in scoliotics patients possibly resulting from the asymmetrical body masses. Due to this fact, the severity of spine deformation can be bound on the asymmetries and postural instability.

Nowadays, the assessment of the Cobb angle is the master key to adapt AIS treatment. This clinical index calculated from a patient's back X-ray allows the measure of the angle of the AIS curvature [33]. Such assessment is limited because only the coronal plane has been taken into account; however, it is easy to use by the clinicians [10, 15, 26]. The correlation between the Cobb angle and the postural parameters is controversial. In standing position, some authors showed an increase of the instability correlated to Cobb angle [1, 26] while Shalstrand et al. [27] did not show such correlation. During gait, the asymmetries computed between lower limbs are not correlated with Cobb angle [10, 21, 29]. Controversial results may be attributed to the chosen processed variables but also to the standing position as experimental paradigm (posture and walking).

Unstable sitting posture associated with added asymmetrical masses compromises the mechanical balance of the trunk and suggests the necessity of adjustments which are directly dependent of the internal mass repartition (AIS) and external forces (added masses) [6, 8]. Such adaptive strategies related to AIS pathology should be variables sensitive to the progression of the spinal deformation. Self-destabilising movement was demonstrated priorly to be relevant in the extraction of the strategies based on the increase of both variability and performance index (PI) with the spinal deformity [6]. While differences between patients and control subjects exists in two movement directions [5, 6], the ML destabilisation induce an increase of the groups differences. However, increasing variability of the parameters in such specific direction may reduce the link between the Cobb angle and the postural variables.

The aim of the present study was therefore to investigate the biomechanical factors involved in the postural responses of seated AIS patients to an angular acceleration of the seesaw in function of severity of spine deformation commonly represented by Cobb evaluation.

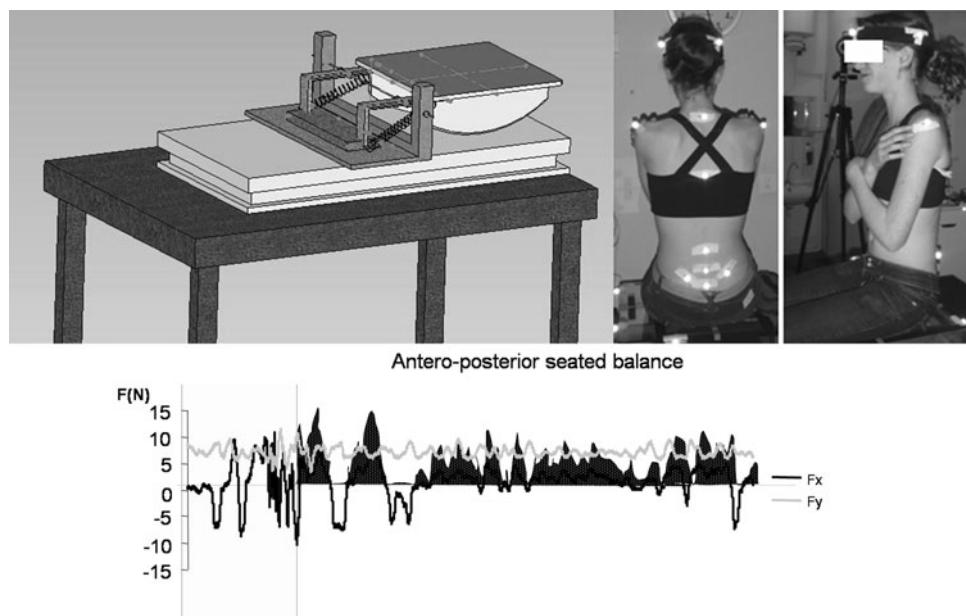
Materials and methods

Twelve female adolescents with right thoracic AIS (without compensatory curves) were included in this study (11.83 ± 0.8 years, height 155.58 ± 4.9 cm and weight 41.92 ± 8.9 kg). This specific curvature implied larger modifications in terms of dynamic control than for the lumbar or compensatory curvature [15]. The patients were recruited at a children's spinal rehabilitation ward by a rehabilitation specialist, who assessed the stage of AIS using Cobb's method (mean angle: $30.17^\circ \pm 9.7$ —range $19\text{--}45^\circ$). The mean gibbosity of the curvature was $10.16^\circ \pm 2.7$, and Risser's test gave a mean value $1.4/5 \pm 0.6$. The patients had no associated pathologies. The only treatment they were undergoing was wearing a corset and physiotherapy (the corset was worn during the daytime and/or at night: it had been worn prior to these experiments for 7 months on average). All the tests were performed without their brace (the patients were asked to leave out the brace minimum 4 h before testing). Once the experimental procedure had been validated and all the subjects (who volunteered) duly informed, they and their legal representatives gave their prior consent in writing.

The experiments were carried out at the department of functional exploration or at school. All the experimental procedures were carried out using the same experimental set-up. Dynamic analyses were performed by recording forces signal obtained from an AMTI® forceplate installed underneath a seesaw placed on a table (Fig. 1). The ground reaction forces (GRF), AP (Fx) and ML (Fy) forces were recorded at 500 Hz acquisition frequency [6, 11]. The seesaw used, specially designed for testing seated subjects' balance, consisted of three parts: a plateau which was affixed to two perpendicular planes with a curved bottom (Fig. 1). Anthropometric measurements were carried out on 72 subjects in order to define the radius of curvature of the 300 mm long seesaw plateau ($r = 0.91$) corresponding to the destabilisation of the spinal segment when the axis of rotation was centred on the T11 vertebra for a subject 150 cm tall. This spinal level corresponded to the last vertebra of the scoliotic curvature for a thoracic scoliosis. Lastly, a mechanical device was developed to maintain the plateau of the seesaw horizontal so as to be able to control the onset of the postural disturbance (t_0) applied to subjects in the sitting position (Fig. 1).

At the beginning of each test, the subject was seated on the seesaw stabilised in the horizontal position using the special external device (Fig. 1) designed to keep it in a fixed position at the beginning of the test and to free it when required. The force signals began to be recorded at the instant (t_0) of the release of the seesaw by the experimenter so that the subject was destabilised in the AP or ML direction. The task consisted of keeping balance for 10 s.

Fig. 1 The experimental set-up and data analysis. The *top panel* shows the experimental set-up, consisting of a force platform, a seesaw and a destabilising device. The *bottom panel* shows the changes with time (ms) in the ground reaction forces (N) recorded (F_x -anteroposterior and F_y -mediolateral). The performance index (PI) was the area under the curve (black)



The experimental conditions were run in randomised order and each condition was repeated three times. The conditions involved unload condition with eyes open (EO), the asymmetrical loading of the shoulder (left shoulder-LS vs. right shoulder-RS) and the pelvis (right pelvis-RP vs. left pelvis-LP). The load added, which corresponded to 15% of the weight of the body segment in question [12], was that found to increase the inertia of the trunk segment in walking for AIS subjects. Vernazza et al. [35] have suggested that this increase in the segmental inertia may be due to a combination between age and parameters such as the body weight, the segment masses and the segment length. A 0.50 kg load was therefore placed in the latter study [14] on the shoulders and 2 kg load on the trunk of 10-year-old subjects (mean mass: 36 kg), and a 0.7 kg load on the shoulders and a 3 kg load on the trunk of 14-year-old subjects (mean mass: 56 kg). To increase the mechanical effects of the asymmetrical loading, masses at least 1 kg heavier than these values were used in the present tests.

The GRF data obtained were processed (MATLABv.6, Mathworks) to determine the variability of the forces (F_x and F_y) during each stabilisation (Fig. 1). Then, a PI was computed ($PI = \text{force} \times \text{duration} = \text{area under the curve}$, Fig. 1). In a previous study [6], it was demonstrated that a self-destabilisation in a seated position created a specific strategy provided by AIS patients compared to a control group. The patients generated a forward displacement oriented towards the concavity of the scoliosis curvature in the frontal plane (left side). Due to that, forward PI and concavity PI (left side) were used to compare AP and ML balance as specific patients' parameters and also to test the correlation between the PI and the Cobb angle.

Statistical analysis aimed: (1) to compare AP and ML self-destabilisations and, (2) to test the correlations between the Cobb angle and, the parameters (variability, concavity PI and forward PI). For each subject, the average of the three tests of each condition was calculated for each parameter. To answer to the first objective, the normality of the population was verified (Shapiro-Wilk test) and, the analysis of the variance (ANOVA) test with its post-hoc effects (Neumann-Keuls test) were run. The answer to the second objective was given through Pearson's correlation test between averaged computed parameters and Cobb angles for the overall patients. A threshold value of $p < 0.05$ was adopted for ruling out the null hypothesis.

Results

For each PI values, the ML destabilisation showed statistically higher values than the AP destabilisation ($p < 0.001$) (Table 1).

The F_x and F_y variability values obtained showed that the ML destabilisation showed also a higher variability ($p < 0.001$) than the AP destabilisation in all conditions tested only for the F_y forces (Table 2).

When the subjects were AP stabilised, the forward PI and the concavity PI were significantly correlated with Cobb angle ($p < 0.05$) for all conditions except for load on the RS for the concavity PI ($r = 0.52$) (Table 3).

Whatever the PI was, the major correlation concerned the EO condition (Figs. 2, 3).

For a AP destabilisation, the F_x parameters (AP forces) showed a strong correlation ($r > 0.63$, $p < 0.05$) between Cobb angle and variability in all conditions (Table 4).

Table 1 Comparison between mediolateral and anteroposterior seated destabilisations for anterior and concavity performance index

Conditions	Mediolateral	Anteroposterior	<i>p</i> value
Anterior performance index (N ms)			
Eyes open	29010.32 ± 9437.98	10886.81 ± 9925.81	***
Left shoulder	32611.88 ± 10596.99	11091.23 ± 9218.98	***
Right shoulder	27304.86 ± 9776.63	9878.53 ± 6769.66	***
Left pelvis	29124.39 ± 9059.33	10265.05 ± 8351.73	***
Right pelvis	29189.64 ± 11803.71	11206.34 ± 9375.22	***
Concavity performance index (N ms)			
Eyes open	10001.12 ± 4896.04	3814.61 ± 2196.54	***
Left shoulder	8513.01 ± 4217.39	4295.47 ± 3439.75	***
Right shoulder	10791.53 ± 6673.36	3635.30 ± 2281.29	***
Left pelvis	13478.80 ± 8406.71	3330.30 ± 1848.16	***
Right pelvis	10416.40 ± 5247.40	3952.02 ± 2481.14	***

Significant differences are accepted for *p* value < 0.05 (*), < 0.03 (**) and < 0.001 (***)[†]. A not significant *p* value > 0.05 is represented by NS

Table 2 Comparison between mediolateral and anteroposterior seated destabilisation for Fy variability values

Conditions	Mediolateral	Anteroposterior	<i>p</i> value
Fy forces			
Eyes open	2.80 ± 1.42	1.08 ± 0.62	***
Left shoulder	2.56 ± 1.19	1.14 ± 0.63	***
Right shoulder	3.06 ± 1.79	1.09 ± 0.62	***
Left pelvis	3.89 ± 2.53	1.05 ± 0.50	***
Right pelvis	3.07 ± 1.46	1.11 ± 0.57	***

Significant differences are accepted for *p* value < 0.05 (*), < 0.03 (**) and < 0.001 (***)[†]. A not significant *p* value > 0.05 is represented by NS

While for Fy parameters, only the RP and LP loading and LS loading showed a similar characteristic ($r > 0.72$, $p < 0.03$). It may be noticed that the RS loading and EO condition are not correlated (non significant results).

Whatever the forces were, the *r* value increased ($r > 0.81$) when the load was added on the convex side (Fig. 4).

For ML destabilisation, no significant correlations were found between Cobb angle and all parameters computed (anterior PI, forward PI, Fx variability and Fy variability, Tables 3, 4).

Discussion

Our results succeed in validating an existing link between the severity of the scoliosis, i.e., the Cobb angle, and the extracted parameters to characterise the postural strategies. The characterisation of the link is possible to describe the correlation between Cobb angle and postural strategies adopted by the AIS subjects with AP or ML perturbation in seated position. The AP destabilisation is the most pertinent condition for linked Cobb angle and GRF parameters. The scoliotic patients were characterised by larger changes in the ML destabilisation compared to AP destabilisation only for ML GRF component and both PI in the forward and leftward directions, corresponding to the concave side of their spinal curvature.

The adaptive spatio-temporal responses to self-destabilisation of a seated position were characterised by the increase of GRF variability and an increase of PI in the forward position and concavity side [6]. These strategies in seated position are, therefore, similar to those obtained during locomotion [16], lateral step [4], gait initiation [5] and in the standing position [8]. These dynamic strategies also constitute an adaptive response to sensorimotor deficits [7, 15, 27] and scoliotic patients' impaired spatial perception during dynamic postural control [9]. If the sensorimotor and proprioceptive information used to perform motor activities is perturbed, this is bound to affect the patients' motor strategies [23, 36]. The spinal deformation associated with AIS, which involves all three spatial planes, results in the redistribution of body masses, as shown by the asymmetrical ischial thrust recorded in these patients in the sitting position [32] and the asymmetrical

Table 3 Correlation between Cobb angle (°) and performance index values (N ms)

	Mediolateral destabilisation		Anteroposterior destabilisation	
	PI anterior	PI concavity	PI anterior	PI concavity
Right pelvis	NS	NS	0.72 (**)	0.58 (*)
Left pelvis	NS	NS	0.62 (**)	0.62 (**)
Right shoulder	NS	NS	0.76 (**)	0.52 (NS)
Left shoulder	NS	NS	0.67 (**)	0.57 (*)
Eyes open	NS	NS	0.88 (***)	0.69 (**)

Significant differences are accepted for *p* value < 0.05 (*), < 0.03 (**) and < 0.001 (***)[†]. A not significant *p* value > 0.05 is represented by NS

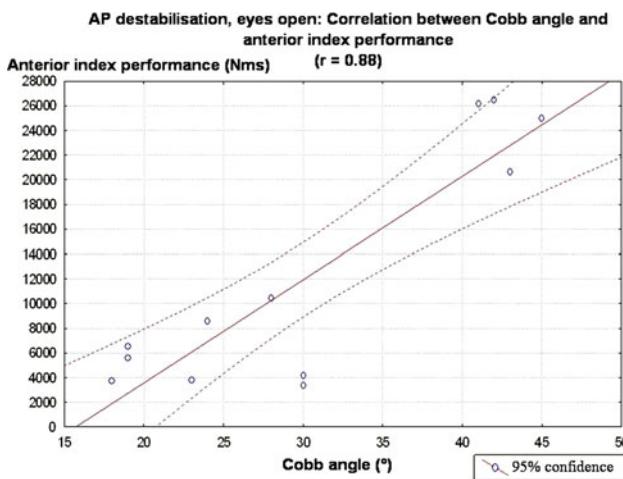


Fig. 2 Correlation between Cobb angle (°) and anterior index performance (N ms) for an anteroposterior destabilisation

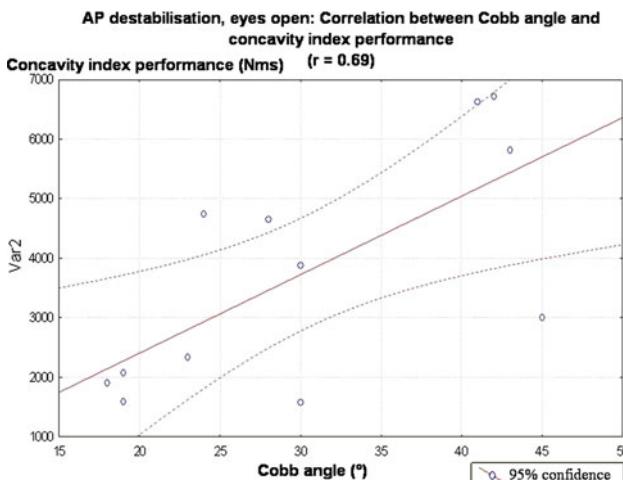


Fig. 3 Correlation between Cobb angle (°) and concavity index performance (N ms) for an anteroposterior destabilisation and the eyes open condition

GRF exerted during the performance of motor tasks [5, 29]. It was therefore proposed here to increase the inertia of the trunk by adding masses in order to establish how inertial parameters affect the patients' dynamic behaviour. Additional masses of this kind did not affect the subjects' balancing strategies when the perturbation was applied in the AP direction [6]. Moreover, the AP destabilisation with additional masses on the pelvis and on the shoulders showed the higher correlations between the Cobb angle and the GRF variability. Thus, the postural imbalance linked with the curve severity was directly affected with a mass modification on the trunk. At the opposite, adding extra masses lowered the correlation between Cobb angle and PI. Thus, moving the trunk forward and towards the side of the concavity corresponded in developing a postural strategy, curve severity dependent to stabilize the upper part of the

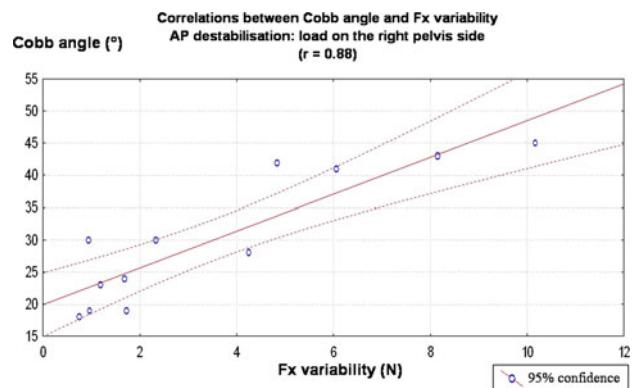


Fig. 4 Correlation between Cobb angle (°) and Fx variability (N) for an anteroposterior destabilisation and the load on the right pelvis side (convexity scoliosis side)

Table 4 Correlations between Cobb angle and Fx and Fy variability values

	Mediolateral destabilisation		Anteroposterior destabilisation	
	Fx	Fy	Fx	Fy
Right pelvis	NS	NS	0.88 (***)	0.81 (**)
Left pelvis	NS	NS	0.8 (**)	0.78 (**)
Right shoulder	NS	NS	0.74 (**)	NS
Left shoulder	NS	NS	0.87 (***)	0.72 (**)
Eyes open	NS	NS	0.63 (*)	NS

Significant differences are accepted for p value < 0.05 (*), < 0.03 (**), and < 0.001 (***). A not significant p value > 0.05 is represented by NS

body. The addition of additional masses faced the subject to a new issue which prevented him from performing the optimal strategy; such point explained the decreased correlation with the Cobb angle due to a less appropriate response [24].

While in the literature, the correlation between the Cobb angle and the postural parameters is controversial [1, 10, 21, 26, 29], in seated position, an high correlation between these parameters was found. Even if such discrepancy may be due to the chosen parameters, the position itself seemed to be more relevant. The standing posture reflected, through the modification of GRF, the global segmental organisation of the body [25, 29]. Contrary to standing, seated posture enlightened the specific postural control of the trunk that allowed the assessment of specific balance strategies [28, 32]. Standing and unstable balance presented similarities as higher variability, larger impulses and larger GRF displacements when facing AIS patients to control subjects [4, 5]. It appeared that the variability of the strategies and the number of involved corporal segments were associated with the movement that determined the

relationship between the Cobb angle and the postural characteristics of the patient.

In a sitting position, balance mainly concerned the head–trunk–upper limbs segments. However, the mass of the lower limbs was still acting forward and would modify the postural adjustments. Certain studies, to limit the effect of the lower limb, proposed different solutions to link lower limb to the seesaw [11, 20, 31]. In these configurations, the lower limbs followed the plateau displacement. However, with such devices no control of the compensation effect due to the lower limbs leaning on a support was possible. Consequently, in our study the lower limbs were set free and the control of the initial stability of the seesaw was provided by a simple mechanical device to make our protocol in clinical environment easily reproducible.

Our results showed that the correlation between the Cobb angle and the postural strategies was plane of progression dependent. Only a sitting anteroposterior imbalance which was perpendicular to the frontal spinal deformity linked both parameters. Winter et al. [37] demonstrated that for control subject, the orthogonal force to the movement reflected the balance control. On the other hand the whole of our studies about the influence of orthogonal movements, ML versus AP, on balance control demonstrated that scoliosis systematically induces a perturbed AP GRF component [5]. In addition the relationship between the scoliotic curve severity and the dynamic perturbations was also remarkable for a AP movement (perpendicular to the frontal deformation). It appeared that to study the influence of the frontal curvature on muscular asymmetry, on ML displacements and on ischiatic support, the sagittal plane is the one of interest (force and AP movement) [17, 28, 32]. The extraction of an index taking into account the tri-dimensionality of the deformity would be differently correlated with the postural parameters. Whereas a movement performed in the direction of the frontal deformity would appear interesting, it induced an increase of the variability of the dynamic parameters [28] and a lesser validity.

In conclusion, the perturbations of the postural organisation which were systematically noticed in the AIS group confirmed the importance to develop a clinical index to identify the corresponding strategies [5, 8, 16]. Stabilometry techniques were already used to assess the postural control in the case of scoliosis [11, 34]. Because of the complexity and the lesser reproducibility of the kinematic versus dynamic parameters [19, 34], the clinical index must be extracted from GRF data. The seated unbalanced position provided relevant paradigm compared to standing. Actually, such self-destabilisation sitting revealed the specific behaviour of the scoliotic patients which is similar to standing in terms of variability and, PI. Self-destabilisation sitting also allows to link the severity of the scoliotic curvature to the relevant dynamic parameters, such as GRF

variability, the concavity side and forward PI. In order to get ready to use such index, next longitudinal studies must be realised to indicate the relevant clinical thresholds and to establish the prognostic factor.

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